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# Predictions of Mechanical Output of the Human M. Triceps Surae on the Basis of Electromyographic Signals: The Role of Stimulation Dynamics

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*In order to assess the significance of the dynamics of neural control signals for the rise time of muscle moment, simulations of isometric and dynamic plantar flexion contractions were performed using electromyographic signals (EMG signals) of m. triceps surae as input. When excitation dynamics of the muscle model was optimized for an M-wave of the medial head of m. gastrocnemius (GM), the model was able to make reasonable predictions of the rise time of muscle moment during voluntary isometric plantar flexion contractions on the basis of voluntary GM EMG signals. The rise time of muscle moment in the model was for the greater part determined by the amplitude of the first EMG burst. For dynamic jumplike movements of the ankle joint, however, no relationship between rise time of muscle moment in the experiment and muscle moment predicted by the model on the basis of GM EMG signals was found. Since rise time of muscle moment varied over a small range for this movement, it cannot be completely excluded that stimulation dynamics plays a role in control of these simple single-joint movements.*

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**Keywords:** Muscle Model, Human Triceps Surae, Electromyographic Signals, Stimulation Dynamics, Excitation Dynamics, Contraction Dynamics.

## Introduction

Muscle models play an important role as actuators in simulation studies that address questions in the field of multisegment movement control. A muscle model typically consists of two sets of equations, one describing the contractile behavior of muscle, the other describing its excitation by the central nervous system. The former is usually called the contraction dynamics of the muscle model; the latter its excitation dynamics. The rate of force development by a muscle model is determined by the combined effect of contraction and excitation dynamics and, third, the rate of change of the control signals to the model, for which the term *stimulation dynamics* will be used. In a previous paper [1], both contraction dynamics and excitation dynamics were determined on the basis of experimental data for the human m. triceps surae (TS) muscle-tendon complex (MTC). It was shown that using these dynamics, a muscle model representing TS is able to predict the rate of muscle moment development correctly for electrically elicited isometric plantar flexion.

The role of stimulation dynamics for voluntary muscle activation has not yet been studied extensively. To perform such a study, a method to measure neural control signals to the muscle would be required. In case of contractions elicited by electrical stimulation, neural control signals are well defined by current pulses applied to a peripheral nerve. For voluntary muscle activation, however, a direct measure of neural control signals is not available. It is only possible to obtain bio-electrical signals from active muscles in the form of electromyographic (EMG) recordings. These EMG recordings represent the averaged electrical activity of a number of motor units in the vicinity of the recording

electrodes. Although this electrical activity is essentially an output of muscle, it is closely related to the neural control signal to the units and has therefore successfully been used as an input for muscle models (e.g., [2,3] and others).

The purpose of this study is to determine the role of stimulation dynamics for muscle moment development in voluntary isometric contractions and single-joint jumplike movements in which TS is used for propulsion, using EMG signals of this muscle group as a measure for neural control signals. To this end, excitation dynamics will be optimized so that moment histories during an electrically elicited isometric twitch of TS are predicted correctly when the M-wave response due to this stimulus is used as input for the muscle model. Subsequently, it will first be evaluated whether this excitation dynamics is capable of predicting mechanical output during voluntary twitchlike contractions of TS using voluntary EMG signals as input. This is not a trivial step, since during voluntary contractions, in addition to changing the firing rate of active motor units, muscle moment can be modulated also by recruitment of new units. Second, using the optimized excitation dynamics, it will be investigated to what extent mechanical output of TS can be predicted for voluntary isometric contractions of TS on the basis of voluntary EMG signals. Finally, the same question will be investigated for dynamic single-joint jumplike movements of the ankle joint.

## Methods

**Subjects.** Three male subjects participated in this study. All subjects had also participated in a previous study [1]. From each subject, informed consent was obtained according to the policy statement of the American College of Sports Medicine. Characteristics of the group were (mean  $\pm$  standard deviation): age  $26 \pm 0.5$  year, height  $1.84 \pm 0.04$  m, body mass  $75 \pm 8$  kg. All subjects participated in two experiments on the calf muscles of their right leg. In the first experiment isometric contractions were re-

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corded using a dynamometer, while in the second experiment dynamic jumplike movements were recorded with the use of, among other things, a force platform.

**Dynamometer Experiments.** In these experiments, both electrically elicited and voluntary isometric contractions of the plantar flexors of the right leg were recorded from each subject on a KinCom isokinetic device (Chattecx Corp., Chattanooga, TN), operating in isometric mode. During measurements, the subject was lying in a prone position, with the knee of his right leg fully extended. The ankle joint was carefully aligned with the axis of rotation of the dynamometer. In all experiments, the ankle was kept in a slightly plantar-flexed position (100 deg, angle between tibia and sole of the foot). Pairs of disposable surface electrodes (Meditrace ECE 1801) were applied to the skin overlying m. soleus (SOL), and the medial head of m. gastrocnemius (GM) after standard skin preparation [4]. In order to assess the amount of co-contraction during voluntary contractions, surface electrodes were also placed on the skin overlying m. tibialis anterior (TA), the principal antagonist of TS. Electrical signals from the muscles were pre-amplified 100 $\times$ , 20 Hz high-pass filtered to eliminate movement artifacts and 1500 Hz low-pass filtered using K-lab MS 101 preamplifiers. Next, EMG signals were further amplified. Off-line, EMG recordings were rectified and low-pass filtered using a second-order zero-lag digital Butterworth filter with a cut-off frequency of 22 Hz. In order to prevent recording of a stimulus artifact during the electrically elicited contractions, EMG amplifiers were blanked for 2 ms after each stimulation pulse using custom-made equipment.

TS was stimulated artificially through the n. tibialis in the popliteal fossa using 100  $\mu$ s current pulses applied to the nerve by means of a pen electrode. For this purpose, a computer-controlled stimulator identical to the one described by Prochazka et al. [5] was used. Electrode position was standardized throughout the experiment by the use of an electrode holder similar to the one described by Simon [6], fixed to the leg by means of elastic bandages. Initially, using a low level of current, the electrode was positioned with respect to n. tibialis such that the largest mechanical response in TS was obtained. Subsequently, the level of current was increased until a twitch of at least 12 Nm was obtained. More details concerning the experimental procedures can be found in van Zandwijk et al. [1]. During all experiments, plantar flexion moment, joint angle, and EMG activity from TS were sampled simultaneously at 1000 Hz and stored for subsequent analysis.

In order to optimize excitation dynamics using EMG signals as inputs, electrically elicited contractions at a stimulation frequency of 1 Hz were recorded from TS. Second, to be able to evaluate performance of this excitation dynamics for voluntary contractions, TS was stimulated at a frequency of 1 Hz and the subject was instructed to perform brief voluntary plantar flexion contractions between two successive electrically elicited twitches, thereby attempting to approximate the mechanical response of the electrically elicited twitch. Visual feedback of moment histories was provided to the subject to facilitate performance of this task. Third, the subject was instructed to perform maximal voluntary contractions (MVCs) of his plantar flexors. Contractions were performed in which it was emphasized to contract as *forcefully* as possible, followed by contractions with the explicit instruction to contract as *quickly* as possible. Finally, the subject was asked to perform a series of isometric contractions at five different levels of steady-state moment, up to the level reached during a MVC. On the basis of these trials, the gain factor between EMG level and muscle moment was determined for each part of TS.

**Force Platform Experiments.** On a separate day, dynamic jumplike movements using the plantar flexors of the right leg for propulsion were performed by each subject. During execution of these movements, positions of retro-reflective markers placed on anatomical landmarks were registered using electronically shut-

tered cameras (NAC 60/200 MOSTV). Markers were placed on the fifth metatarsophalangeal joint, lateral malleolus, knee joint (on the lateral collateral ligament at the height of the joint cleft), greater trochanter and trunk. Simultaneously, vertical and fore-aft components of the ground reaction force were recorded using a force platform (Kistler 9281B, Kistler Instruments Corp., Amherst, NY). EMG electrodes (Meditrace ECE 1801) were placed on the same muscles as described in the previous section. Electrical signals of the muscles were amplified (Disa 15 C01, Disa Electronics, Skovlunde, Denmark) and 7 Hz high-pass filtered to eliminate movement artifacts. Subsequently the electrical signals were rectified and 22 Hz low-pass filtered. Marker coordinates and EMG signals were sampled at a frequency of 200 Hz. Afterward, net joint moments were calculated using a standard inverse-dynamics approach.

First, the subject performed one-legged jumplike movements of the ankle joint. The subject started the movement with the foot flat on the ground and was instructed to keep his knee and hip joints fully extended during execution of the jump and to jump without making any preparatory countermovement. Next, the subject performed a quasi-isometric contraction of the plantar flexors of his right leg. In this contraction, the subject was instructed to increase the moment exerted by the plantar flexors by attempting to rise slowly on his toes and then to decrease the moment again slowly. During this contraction, he restrained himself by holding on to a bar fixed to the floor by means of a chain, thereby preventing extension of the ankle joints. On the basis of this trial, the gain factor between EMG level and ankle joint moment was determined for each part of TS.

**Normalization of EMG Signals.** In the dynamometer experiments, M-waves recorded from TS were divided by their peak amplitude before they were used as inputs for the muscle model. EMG signals recorded during both the alternated electrically elicited and voluntary twitchlike contractions and the MVCs of TS were normalized by dividing them by the same M-wave peak amplitude, yielding normalized EMG (NEMG) signals. In order to relate EMG signals recorded in the force platform experiments to those recorded in the dynamometer experiments, the following calibration procedure was adopted. First, EMG signals recorded in the force platform experiment were multiplied by the ratio of gain factors as determined in the dynamometer and force platform experiment. Next, NEMG signals from each part of TS were obtained by dividing EMG signals by the peak amplitude of the M-wave recorded in that part.

**Muscle Model.** Simulations of isometric and dynamic muscle contractions were performed using a Hill-type muscle model, which has been described extensively elsewhere [1,7-9]. The key features of the muscle model are briefly described in the appendix. Input to the model is a NEMG history recorded from TS; output is muscle moment. Excitation of muscle is modeled as described by Hatze [10,11]. This excitation dynamics was originally derived for the use with sine-shaped stimulation pulses  $\text{stim}(t)$  as input. Using NEMG histories as input instead means that besides firing frequency, the *amplitude* of the control signals to the muscle model can also change. When the muscle fibers of each motor unit are distributed uniformly throughout the muscle and the recording electrodes sample a large volume of the muscle, any increase in NEMG signal due to recruitment of an additional unit will be proportional to the size of that unit. So when differences in firing rate between active units are not too large and when there is not a great deal of overlap between unit NEMG potentials, the NEMG signal will represent a measure of the relative amount of the muscle being activated. Under these assumptions, NEMG signals can therefore be viewed as a measure for the *average* stimulation of muscle.

The muscle model consists of a set of five coupled ordinary differential equations solved by numerical integration, using a variable order-variable step size integrator [12]. Since integrator

step size is not constant, it can occur that muscle stimulation needs to be known at instants at which no NEMG data are available. In those cases, muscle stimulation is obtained by means of cubic spline interpolation of NEMG histories.

**Numerical Experiments.** Parameter values describing the contractile behavior of TS have been determined for all subjects in a separate experiment on the basis of a set of isokinetic release contractions [1] using a technique evaluated for rat muscle [9]. In order to perform simulations of muscle contractions using NEMG histories as input, excitation dynamics was first optimized such that the muscle model correctly predicts the mechanical response during an isometric twitch elicited by electrical stimulation of the tibial nerve, using a normalized M-wave recorded from TS as input (see also [1,8]). The numerical techniques used to perform this optimization have been described in full detail elsewhere [1,8]. Subsequently, performance of this excitation dynamics was evaluated by feeding NEMG histories recorded during voluntary isometric contractions of TS into the muscle model and comparing mechanical response of the model to the experimentally recorded muscle moment. Differences between experimentally recorded muscle moment and muscle moment predicted by the model were quantified by calculating normalized root mean square (rms) errors  $\varepsilon$ :

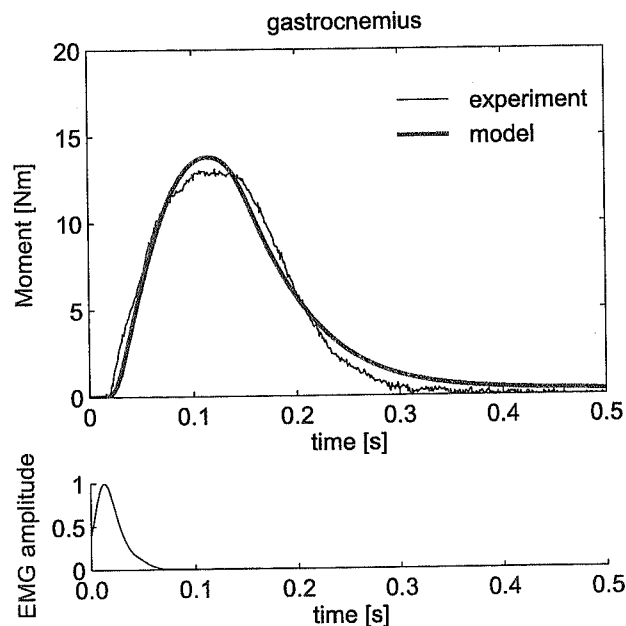
$$\varepsilon = \sqrt{\int_{t_1}^{t_2} (M_e - M_m)^2 dt} / \sqrt{\int_{t_1}^{t_2} (M_e)^2 dt} \quad (1)$$

Here,  $M_e$  is muscle moment obtained from the experimental data,  $M_m$  muscle moment predicted by the model, and  $t_1$  and  $t_2$  are the start and end times of the contraction, respectively. To quantify the performance of the model in predicting the rate of change of muscle moment, rise times (RTs) of variables were calculated. The RT of a variable is the time required for that variable to rise from 10 to 90 percent of its maximal value.

Subsequently, the muscle model was used to predict mechanical output of TS during dynamic jumplike movements propelled by the plantar flexors. For this purpose, the same contraction dynamics as in simulations of the dynamometer experiments was used, except for the maximal isometric moment, which was adjusted to match the peak moments encountered in the jumplike experiments. Excitation dynamics was also the same as obtained by optimization of model behavior for an electrically elicited isometric twitch in the dynamometer experiments. For predicting mechanical output of TS in this case, a combination of forward and inverse dynamics was used. In such an approach, both NEMG histories recorded from TS and TS MTC length histories were provided as inputs to the model and TS muscle moment was calculated by the model. Next, muscle moments predicted by the model were compared to those from kinetic analysis of the jumplike movements by calculating normalized rms errors  $\varepsilon$  during the push-off phase. Similar approaches have been used to predict muscle moments during walking and stepping [13], jumping [14], and sprint running [15].

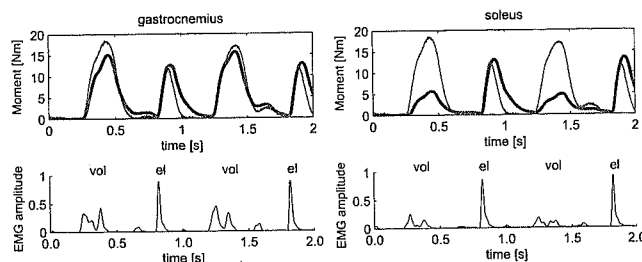
## Results

**Dynamometer Experiments.** Figure 1 gives a typical example of model predictions of moment histories recorded during an electrically elicited isometric twitch contraction. Shown are experimentally recorded moment histories as well as predictions of muscle moment by the model after excitation dynamics has been optimized using a normalized M-wave recorded from GM as input, shown in the lower panel. Subsequently, performance of this excitation dynamics in predicting isometric muscle moment was evaluated for alternating electrically elicited and brief voluntary contractions of TS. The left-hand side of Fig. 2 shows for the same subject an example of the results of such simulations. The figure shows electrically elicited contractions of TS, alternated with brief voluntary contractions. Note the differences in NEMG amplitude and burst duration between the two types of contrac-



**Fig. 1** Moment histories during isometric twitches elicited in m. triceps surae (TS) by means of electrical stimulation of the tibial nerve for subject 3. The thin line pertains to the experimentally recorded twitch, while the thick line pertains to the twitch generated by the model. The simulated twitch is obtained by optimization of parameters pertaining to the excitation dynamics, using the normalized M-wave of m. gastrocnemius medialis (GM) as input. This normalized GM M-wave is shown in the lower panel.

tion. Nevertheless, the muscle model makes predictions of isometric TS muscle moment on the basis of NEMG histories of GM, which correspond closely to the experimentally recorded ones. The ratio of voluntary to electrical NEMG response during these contractions varied considerably across TS. This is illustrated by the right-hand side of Fig. 2, which shows predictions of muscle moment using SOL NEMG signals as inputs, after excitation dynamics had been optimized for a normalized M-wave recorded from SOL. As can be seen, moment generated in the voluntary

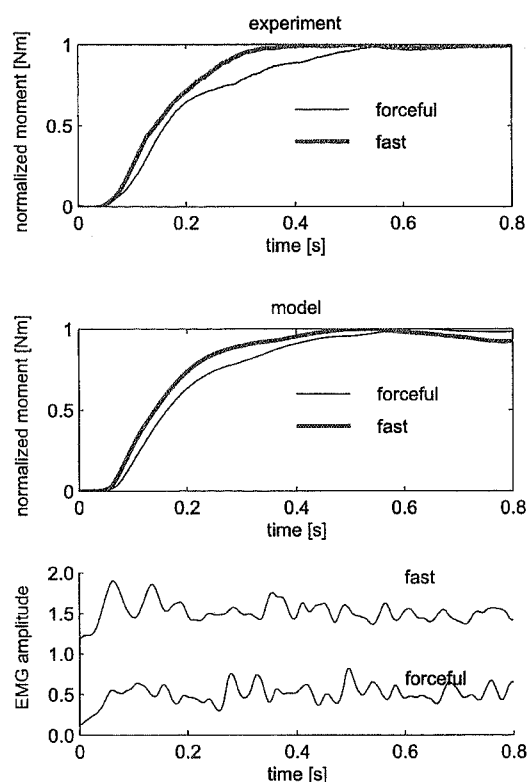


**Fig. 2** Predictions of moment histories for electrically elicited and voluntary twitches of TS on the basis of normalized electromyographic signals (NEMG signals) recorded from m. gastrocnemius (left-hand side) and m. soleus (right-hand side). In both upper panels, the thin line is experimentally recorded muscle moment, the thick line muscle moment predicted by the model. Simulated moment histories are obtained using parameters for the excitation dynamics that have been obtained by optimization of model behavior for an electrically elicited isometric twitch (i.e., the data shown in Fig. 1 for m. gastrocnemius). The corresponding NEMG histories used as input for the model are shown in the lower panels. Electrically elicited twitches are indicated by "el," voluntary twitches by "vol." Same subject as in Fig. 1. For the left-hand side, the normalized rms error  $\varepsilon = 0.28$ , for the right-hand side,  $\varepsilon = 0.70$ .

**Table 1** Normalized rms errors, i.e., the difference between experimentally recorded muscle moment and muscle moment predicted by the model on the basis of NEMG signals as input for alternated electrically elicited and voluntary twitch contractions of TS. For each subject, the values given are the average of three trials.

Subject	Normalized rms error		
	Gastrocnemius	soleus	averaged soleus + gastrocnemius
1	0.71	0.50	0.28
2	0.38	0.61	0.69
3	0.40	0.72	0.51
Average	0.49	0.61	0.50

twitch contractions is largely underestimated in this case. Next, it was investigated whether predictions of experimental muscle moment could be improved by using *averaged* NEMG signals of GM and SOL as input. For this purpose, excitation dynamics was optimized for an isometric twitch, now using the average of normalized GM and SOL M-waves as input. In only one of the three subjects were the subsequent predictions of experimentally recorded muscle moments on the basis of averaged GM and SOL

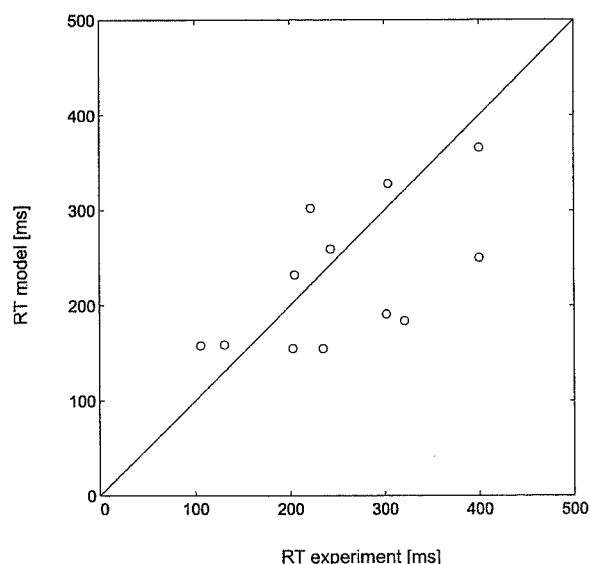


**Fig. 3** Top panel: Experimentally recorded isometric moment histories during maximal voluntary contractions of TS. Middle panel: Predictions of moment histories by the model of TS, using voluntary NEMG signals from GM as input. For both panels, the thin line pertains to a contraction in which the subject was instructed to contract TS as *forcefully* as possible, while the thick line pertains to a contraction in which the subject was asked to contract TS as *quickly* as possible. Moment is normalized with respect to the isometric value. Note the difference in moment rise time between the two contractions in both the experiment and the model calculations. Lower panel: NEMG histories of GM used as inputs for the simulations shown in the middle panel. Note the increased NEMG amplitude at the onset of the contraction in case of the fast contraction. The upper NEMG trace has been shifted upward 1.0 unit to facilitate comparison between traces. Same subject as in Figs. 1 and 2.

NEMG signals better than those obtained using only GM NEMG histories as input. These results are summarized in Table 1, which gives averaged values of the normalized rms error  $\epsilon$  for all subjects. Note the large errors for the predictions of muscle moment histories using SOL NEMGS for subject 3, which corresponds to the data shown in Fig. 2.

Next, performance of this excitation dynamics in predicting the rate of change of muscle moment during MVCs was investigated. The top panel of Fig. 3 gives an example of moment histories recorded during two voluntary isometric plantar-flexion contractions, one being maximally forceful, the other being maximally fast. To facilitate comparison between traces, moment is normalized with respect to its isometric value. Note the differences in rate of change of muscle moment between the two contractions. The middle panel of Fig. 3 shows the corresponding predictions of muscle moment by the model using NEMG signals of GM recorded during these trials as input. Again, muscle moment is normalized with respect to its isometric value. Absolute values of isometric moment were lower than encountered in the experimental recordings. NEMG histories used as inputs for these simulations are shown in the lower panel of Fig. 3. Note the increased amplitude of the first NEMG burst in case of the fast contraction. On the basis of GM NEMG histories only, the model predicts a higher rate of moment development for the maximally fast contraction than for the maximally forceful contraction. During all MVCs of TS, a small amount of activity was recorded in TA, its principal antagonist. Averaged over all three subjects, this activity amounted to 10 percent of the EMG level attained by TA during a maximal voluntary *dorsiflexion* contraction.

Figure 4 shows for all subjects a scatter plot of RT of moment as calculated from the experimental data and RT of moment in the model on the basis of GM NEMG signals. The correlation coefficient between the two quantities was moderate, but significantly different from zero ( $r=0.59$ ,  $p<0.05$ ,  $n=12$ ). Using either SOL NEMG signals or averaged NEMG signals from GM and SOL as inputs for the model yielded correlation coefficients of  $r=0.23$  ( $n=12$ ) and  $r=0.58$  ( $n=12$ ) respectively between RT of experimentally recorded moment and the RT of moment in the model using NEMG signals as input. Further analysis of the GM NEMG recordings revealed a correlation between the amplitude of the first burst and the RT of moment in the model of  $r=-0.69$  ( $n$



**Fig. 4** Scatter plot of rise times (RTs) of isometric moment histories during MVCs of TS for all subjects versus RTs of moment in the model, using GM NEMG signals as input. For each subject, excitation dynamics is optimized for a GM M-wave and contraction dynamics is determined in a separate experiment.

=12) and a correlation between the RT of the GM NEMG signals and the RT of moment in the model of  $r=0.57$  ( $n=12$ ). Also for SOL NEMG and averaged SOL and GM NEMG signals the RT of moment in the model was correlated more strongly with the amplitude of the first burst than to the RT of NEMG recordings in this burst. Therefore, in case of isometric plantar flexion, stimulation dynamics appears to consist in the first place of an *amplitude* modulation of the first NEMG bursts.

**Force Plate Experiments.** Using the same excitation dynamics as obtained from the dynamometer experiments, ankle joint moment was predicted for dynamic jumplike movements of the ankle joint. In this case, inputs to the muscle model were histories of NEMG signals and MTC length and output was muscle moment. In order to obtain NEMG histories, EMG signals were scaled according to the ratio of gain factors found in the dynamometer and force platform experiment, divided by M-wave peak amplitude and subsequently used as inputs for the muscle model. Figure 5 gives an example of histories of muscle moment, GM NEMG signals, and TS MTC length changes obtained during a one-legged jump, using only TS for propulsion. Note the isometric phase at the onset of the movement, where muscle moment is rapidly increasing, while TS has not started shortening. In the top panel predictions are shown of muscle moment by the model on the basis of GM NEMG signals and MTC length changes. Similar results were obtained when SOL NEMG signals and averaged GM and SOL NEMG signals were used as inputs. Table 2 presents values of the normalized rms error  $\varepsilon$  during the push-off phase, as obtained in these calculations for each subject averaged over four

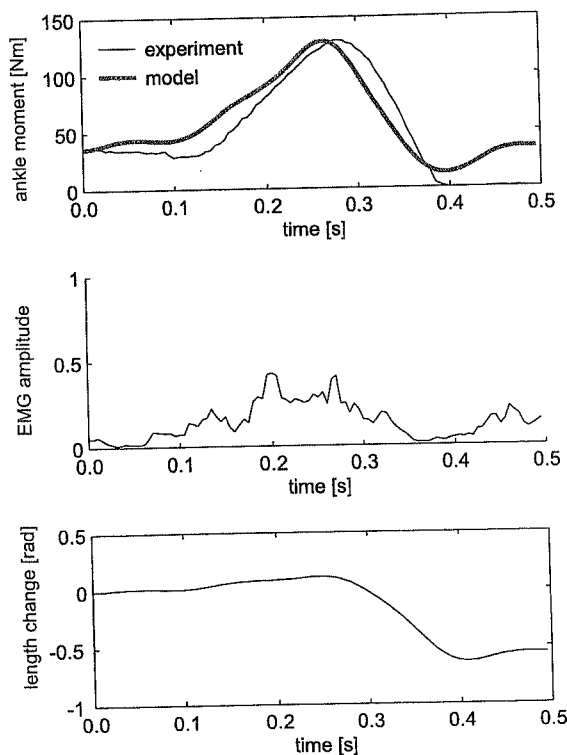
**Table 2 Normalized rms errors between muscle moment predicted on the basis of GM NEMG signals and muscle moment obtained from kinetic analysis for dynamic jumplike movement of the ankle joint using only TS for propulsion. For each subject, the values given are the average of four trials.**

Subject	Normalized rms error		
	Gastrocnemius	soleus	averaged gastrocnemius+soleus
1	0.43	0.34	0.38
2	0.38	0.38	0.40
3	0.28	0.28	0.26
Average	0.37	0.33	0.34

trials. No significant correlations were found between the RT of the experimental moment and the RT of the moment predicted by the model on the basis of NEMG histories.

## Discussion

In this study we set out to determine the significance of the dynamics of neural control signals for moment development. For this purpose, NEMG histories recorded from TS were used as input for a forward dynamic model of this muscle group. To this end, a number of assumptions concerning the behavior of human motor units during voluntary contractions were made. In literature on human muscles, a number of studies supporting these assumptions can be found. First, it was assumed that muscle fibers of each motor unit are uniformly distributed through the muscle. In a histochemical study on a large number of human muscles, Johnson et al. [16] found that the proportion of Types I and II muscle fibers did not vary throughout both m. soleus and m. gastrocnemius. This in contrast to, e.g., m. vastus medialis, where a larger proportion of Type II fibers was found in the surface area than in the deep area of the muscle. In the same study it was reported that muscle fibers of both types were randomly dispersed throughout the cross-sectional area of both m. soleus and m. gastrocnemius. Therefore, the distribution of motor units appears to be uniform throughout TS. Second, it was assumed that the difference in firing rate between active units is not too large. In a study on motor units in m. biceps brachii, m. adductor pollicis, and m. soleus, Bellemare et al. [17] reported frequency histograms for these muscles in which firing frequency for most of the units varied only by factor of about two during a MVC. The smallest range of firing frequencies was reported for m. soleus. Third, for the first dorsal interosseus muscle in humans, Milner-Brown and Stein [18] found that unit EMG potentials to some extent occurred as discrete events, even at relatively high force levels, supporting the assumption that, at least for that muscle, there is no significant overlap between unit EMG potentials. For TS, however, to our knowledge no data concerning this assumption are currently available. In the literature, other models for force regulation during voluntary contractions have been proposed. The model of Fuglevand et al. [19] incorporates individual motor units and generates EMG and force histories on the basis of an independent input. They found relationships between average EMG level and independent input that saturated at high levels of excitation, whereas the model presented in this study assumes a linear relationship. The results presented in Figs. 1 and 2 suggest that GM EMG represents an sufficient measure for muscle excitation during brief electrically elicited and voluntary isometric contractions. For MVCs of TS, however, it might be important to incorporate such nonlinear effects in the model, since in this case muscle moment was underestimated by the model. During isometric plantar-flexion contractions of TS, the EMG level of m. tibialis anterior, its principal antagonist, amounted to 10 percent of its maximal value. This indicates that for these contractions, the amount of co-contraction is small. Together with the fact that the cross-sectional area of m. tibialis anterior is much smaller than that of TS, this gives confidence that net ankle moment is gener-



**Fig. 5 Example of time histories of variables obtained during execution of jumplike movements of the ankle joint. Top panel: Ankle joint moment. Thin line: joint moment derived from inverse dynamic analysis, thick line: moment predicted by the muscle model on the basis of GM NEMG signals and TS MTC length histories as inputs. For this trial, the normalized rms error  $\varepsilon=0.28$ . Middle panel: GM NEMG signals, expressed as fraction of peak M-wave amplitude. Bottom panel: length changes of TS MTC, expressed in angular coordinates.**